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EFFECTS OF LIMITED MOVEMENT ON THE IMPEDANCE PLETHYSMOGRAPH SIGNAL: A PRELIMINARY STUDY

JOHN W. YATES, First Lieutenant, USAF

USAF School of Aerospace Medicine
Aerospace Medical Division (AFSC)
Brooks Air Force Base, Texas

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FOREWORD

This work was done in the Biomedical Engineering Branch under task No. 632001. The study was accomplished in the period from February 1969 to May 1970. The paper was submitted for publication on 10 July 1970.

The author thanks Dr. Harry M. Hughes and Captain Thomas E. Simondi for their many useful suggestions related to the analog and digital analysis of the data.

This report has been reviewed and is approved.

Jo M Quashnock

JOSEPH M. QUASHNOCK
Colonel, USAF, MC
Commander

ABSTRACT

The impedance plethysmograph was evaluated as a means of monitoring respiration of personnel in a space-flight environment. Emphasis was placed on studying the effects of limited movement (tapping of electrodes and skin stretching) on the impedance plethysmograph signal.

A bridge circuit and servospirometer were used to make extremely sensitive measurements of body impedance change and tidal volume. The linear relationship between tidal volume and thoracic impedance was determined by analog and digital methods. These methods were also used to investigate the amount of interference caused by various body movements on the impedance signal. Possible ways of eliminating error from such artifacts were also studied.

Very high linear correlation (r = .85 to .95) was obtained using a bridge circuit. The effects of artifact were reduced by the use of analog filtering, digital programming technics, and a careful choice of display.

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I. INTRODUCTION

With long-duration space flights now a reality, a need exists to monitor respiratory minute volume in a way that will not hinder the astronaut. The impedance plethysmograph used by NASA in the Mercury and Gemini programs gave only qualitative indications of tidal volume. Several methods have been used to measure respiratory changes: the mercury chest band, the spirometer, the nose-clip thermistor, the pressure transducer, and various types of plethysmographs. Although most of these devices gave an indication of respiratory motion in certain circumstances, they were found unsuitable for space-flight application because they restricted the subject, endangered the environment, or did not function properly (53).

Several investigators have used, and reported on, the impedance plethysmograph as a means of measuring respiration. Two basic systems have been used: one based on a bridge circuit (23, 32, 35); the other based on a constant-current source (4, 19, 26, 38). An excellent discussion on each of these systems can be found in the report of Pacela (44).

Although all of the impedance measuring systems described in the literature had disadvantages, two redeeming features made them worthy of further consideration: (1) They all displayed signals that followed respiratory movement regardless of cause. (2) They entailed no hindrance to the body other than that incurred by the electrodes—a parameter already established as necessary for continuous monitoring of ECG. For these reasons the impedance plethysmograph was studied for its adaptability to space-flight monitoring.

In most reported research, the subject was required to be in a supine position with little or no movement, resulting in data that approached the ideal case. Since movement is required in space flight, the aim of this investigation was to study the effect of limited movement on the impedance signal and possible ways to overcome these unwanted signals.

II. METHODS AND MATERIALS

The impedance plethysmograph used for this study was based on the bridge circuit. This instrument was used because, when operated around its null point, it is the most sensitive to impedance changes of all the devices studied. Care was taken to keep within a small range around the null portion of the bridge to eliminate inherent nonlinearities. A parallel resistor and capacitor - balancing network was built using standard Cornell Dubilier decade capacitor boxes and Shallcross resistor boxes. This network, which had values as low as 0.1 ohm resistance and 10 pf. capacitance, permitted an almost perfect initial balance of both phase and impedance.

The electrodes (fig. 1) were built at the USAF School of Aerospace Medicine. These electrodes, which use a coil for the metal portion, are more compact than standard ECG electrodes, yet have a greater surface area for conductivity. The electrodes were placed at

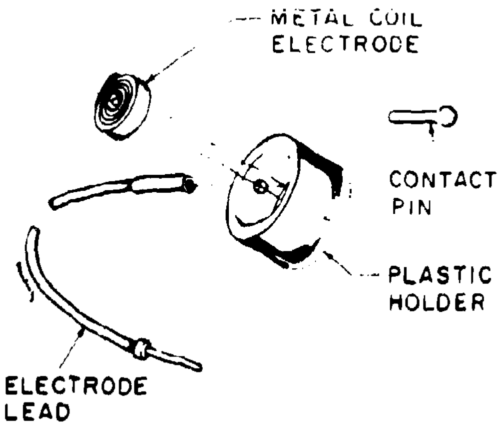


FIGURE 1

ECG recording electrode developed by the USAF School of Aerospace Medicine.

the sixth intercostal space along the midaxillary line. This location was established by Geddes et al. (19) as the position that gave the largest deflection of impedance signal for a respiratory movement. Preparation of the skin area consisted only of cleaning with acetone. The electrodes were then fastened in place with doubled-sided adhesive and electrode paste and allowed to stabilize for a half hour before any measurements were taken.

Respiratory volume was recorded by having the subject breathe into a Servo-Spirometer (Med-Science Electronics, model 350). The spirometer output was calibrated with zero volts, representing the 5-liter mark. Therefore, the lung volumes calculated in this study reflect the volume around the arbitrary value of 5 liters rather than the actual amount of air in the subject's lungs. Since this spirometer was a closed system, the amount of pure oxygen fed into the chamber was controlled by a relay. The relay was activated by a voltage proportional to the volume output of the spirometer. Also recorded for possible comparisons were flow, heart rate, ECG, and impedance phase angle. All of the data were recorded on 1-inch magnetic tape by use of a tape recorder (Ampex FR 1800) and printed out on paper with a Dynograph recorder (Beckman type SII). Figure 2 shows a block diagram of the measuring system used for the experiments.

Three subjects were used for this experiment. Each was required to follow a breathing

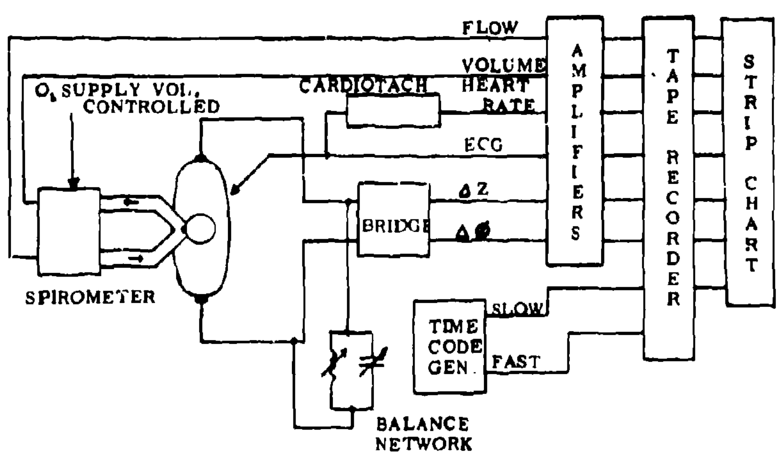


FIGURE 2

Experimental measuring system.

TABLE I

Physical statistics of subjects

Subject	Age	Height	Weight (lb.)	Chest normal	Chest expanded	Waist
1	35	5'11"	165	40"	41"	33"
2	24	5'9"	170	41"	42"	35"
3	35	5'9"	180	40"	41"	35"

pattern that included normal breathing and limited exercise. The physical characteristics of each subject are listed in table I. Table II lists the sequence of breaths and exercises that each subject followed.

X-Y plots were made to investigate relationships between the tidal volume and thoracic impedance. These plots were obtained by simultaneous reproduction of the impedance and volume waveforms on the x and y channels of a Variplotter (E.A.I. model 1110).

Analog-to-digital conversion, needed for the digital analysis, was obtained using an Oscar K digitizer. This machine manually digitizes an analog record and simultaneously punches the numeric values on computer cards. The least-square regression lines of lung volume on thoracic impedance and the correlation coefficients were then calculated for the first normal breathing pattern of each subject. A sampling rate of 10 samples per second was used for this analysis.

Consecutive maximum and minimum points for volume and impedance were then obtained with the digitizer previously cited. The differences between the maximum and minimum points were calculated for the volume and impedance signals. These are defined as ΔV and ΔZ, respectively. Scatter diagrams were plotted of the maximum and minimum points of the volume and impedance waveforms. Scatter diagrams of ΔV versus ΔZ were also plotted.

The slopes obtained from the linear regression analysis, maximum and minimum scatter diagrams, and ΔV versus ΔZ scatter diagrams were compared to determine the accuracy of these methods for predicting tidal volume changes from thoracic impedance changes. The first comparison was made for three subjects. The slope for the linear regression method was used as the standard in this comparison.

TABLE II

Experimental breathing sequence

1. Normal breathing
2. Inhale and hold
3. Normal breathing
4. Exhale and hold
5. Normal breathing
6. Impulse breathing
7. Normal breathing
8. Step breathing
9. Normal breathing
10. Stretching skin near electrodes
11. Normal breathing
12. Tapping electrodes
13. Normal breathing
14. Arm movement up at sides of body
15. Normal breathing
16. Arm movement up in front of body
17. Normal breathing

A second comparison was made to study the relationship between the slopes of normal breathing and artifact data for all three subjects. Slopes from the maximum and minimum plots and ΔV versus ΔZ plots were used in this comparison. This study used the slopes obtained for normal breathing as the standard of comparison.

III. RESULTS

Figures 3 and 4 show the analog x-y plots of various breathing sequences used in this

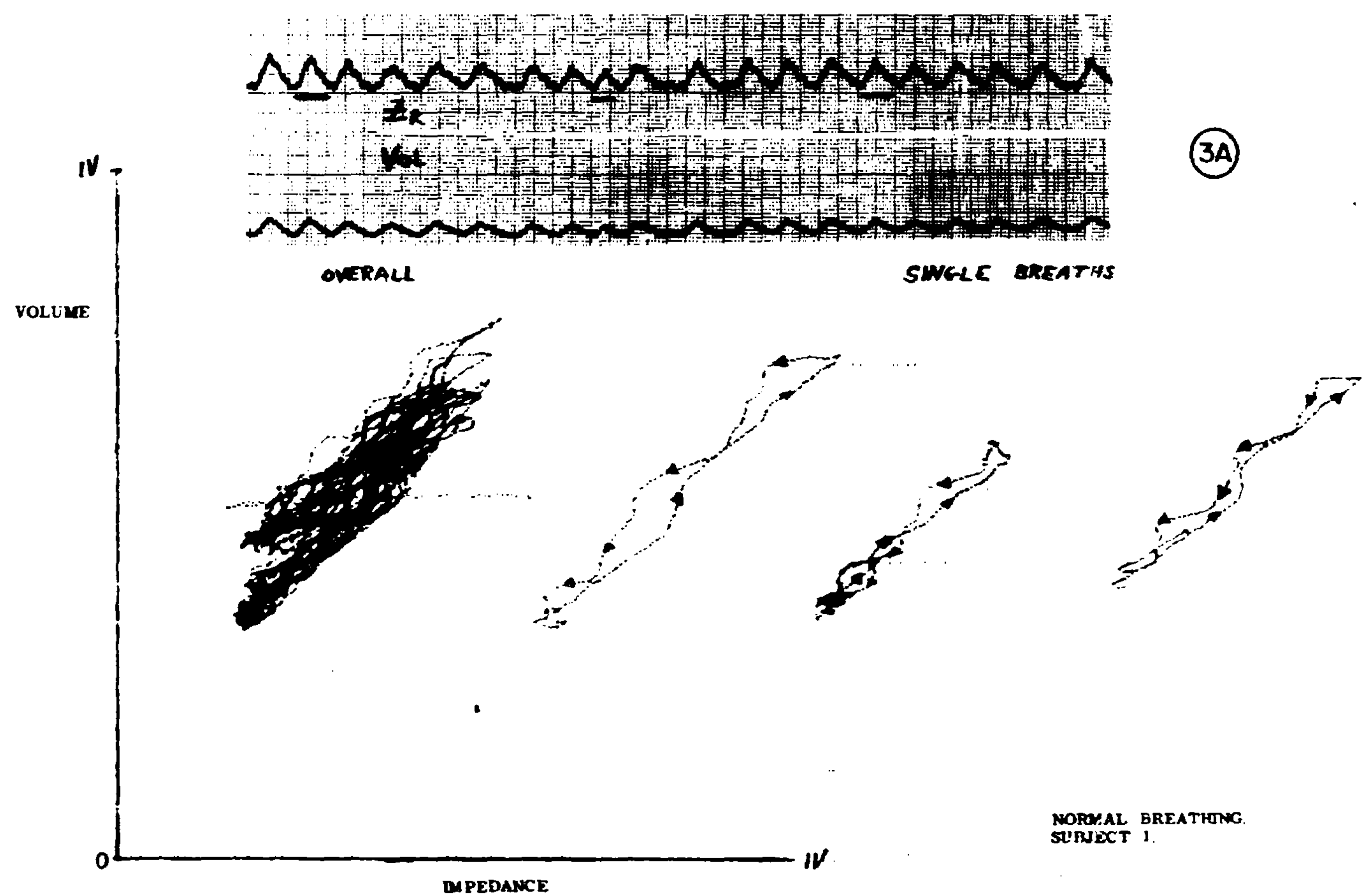
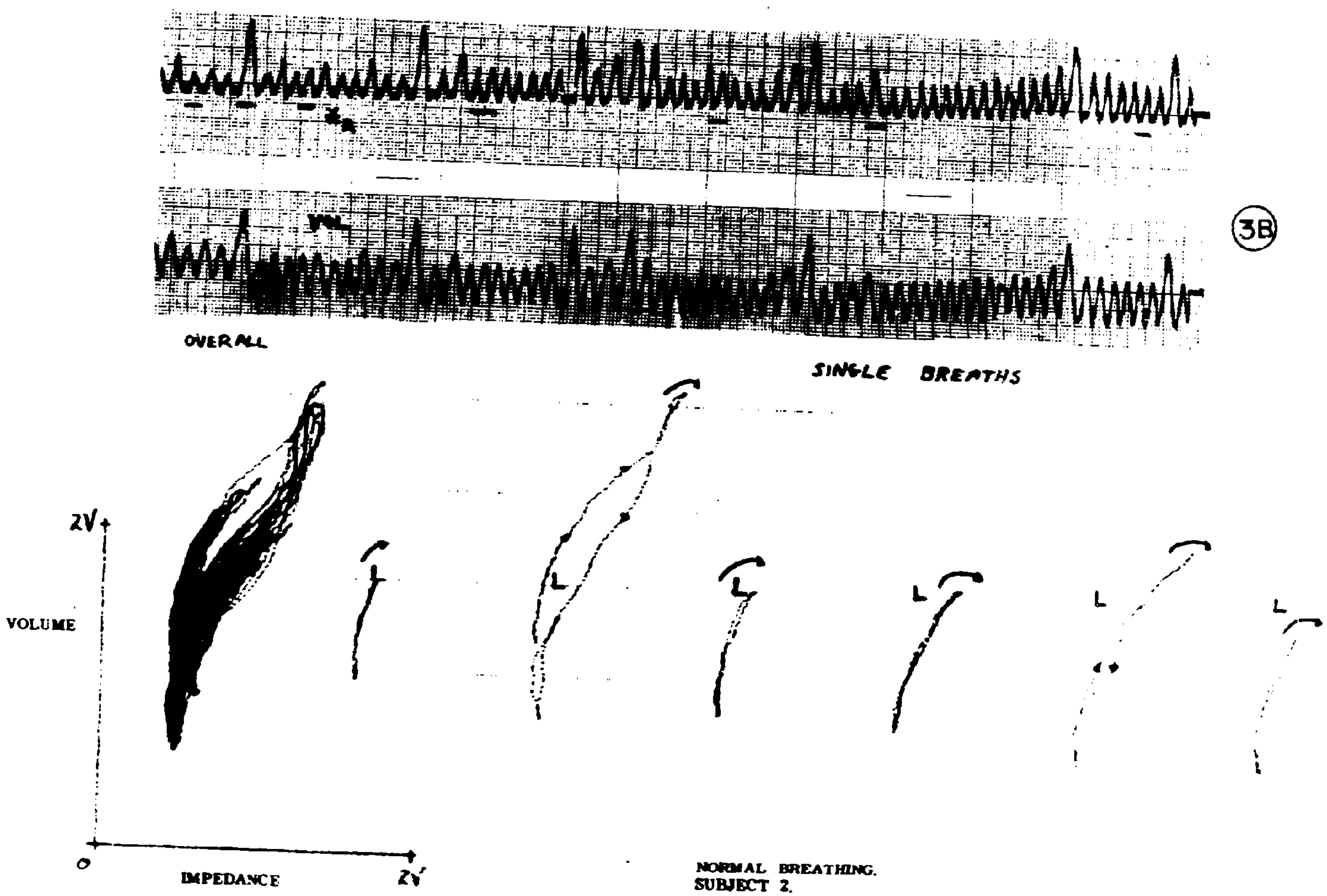
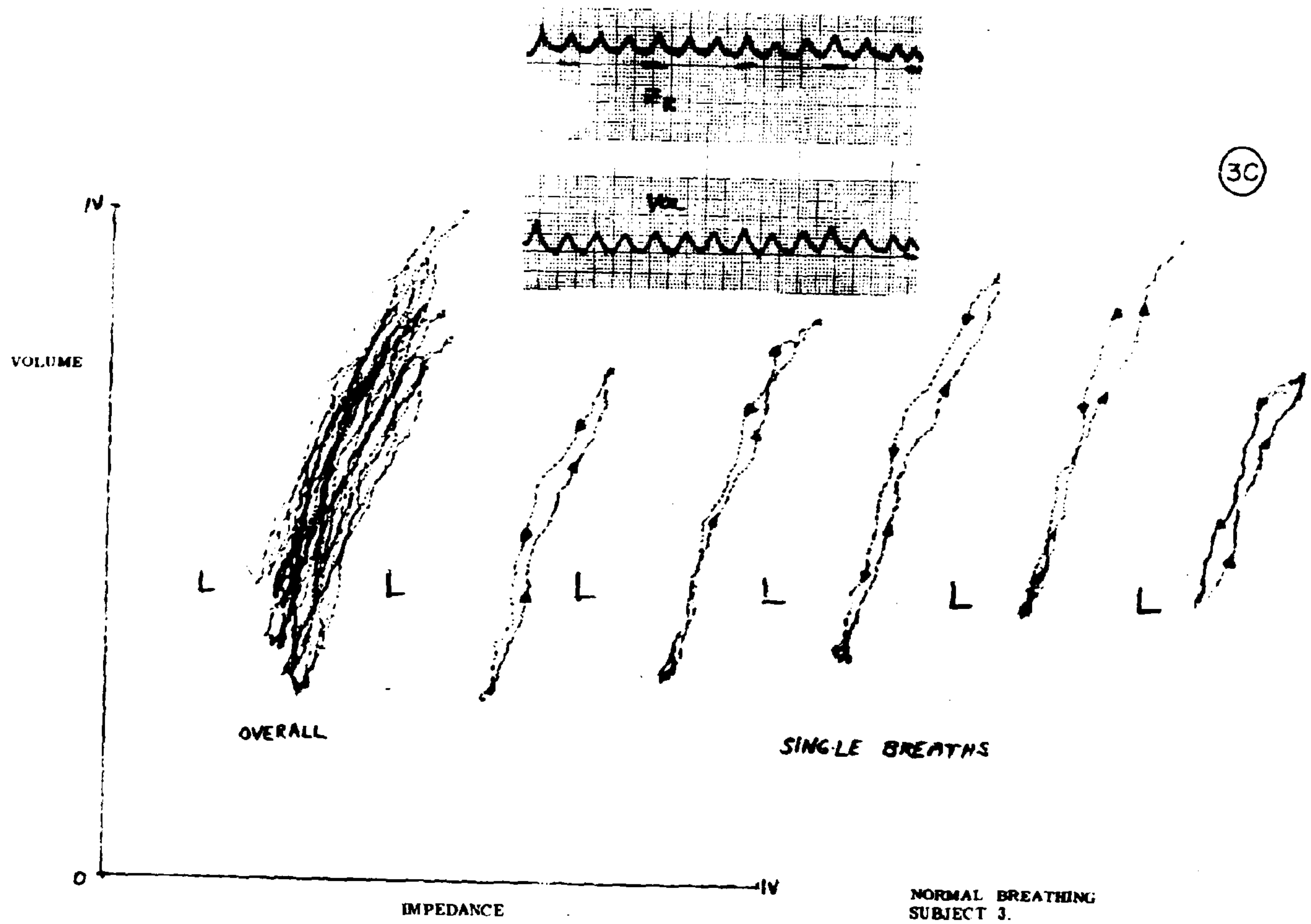


FIGURE 3

Analog x-y plots of normal breathing of three subjects: 3A (subject 1), 3B (subject 2) and 3C (subject 3) are on subsequent pages



NORMAL BREATHING.
SUBJECT 2.



NORMAL BREATHING
SUBJECT 3.

6

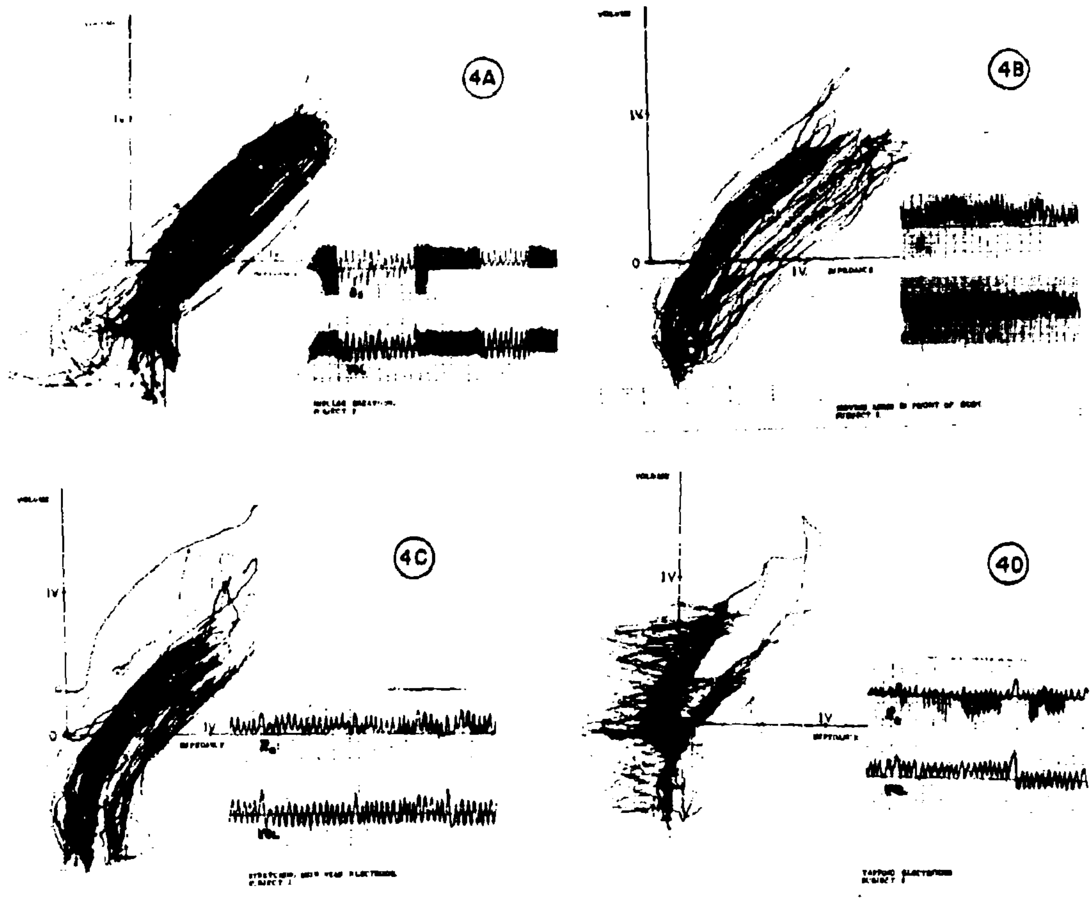


FIGURE 4

Analog x-y plots of various breathing patterns (subject 2): figure 4A—impulse breathing, figure 4B—arm movement; figure 4C—skin stretching; and figure 4D—electrode tapping.

experiment. As indicated on the plots, impedance and volume are represented on the x and y axes, respectively. These plots, as well as all other analog x-y plots, use voltage instead of ohms and liters for the x and y axes. Figure 3 shows the analog x-y plots of normal breathing for all three subjects. The graphs show the pattern of a complete sequence of normal breathing and the waveform of individual breaths. The scale on subject 2 was half that used for the other subjects to accommodate the large excursions made by the subject during normal breathing. Typical x-y plots of impulse breathing, arm movement, skin stretching, and electrode tapping are shown in figure 4. Although there are shifts in the waveforms, a general trend is easily observed because of the large number of breaths recorded.

Least-square regression lines of volume on impedance were then calculated for the normal breathing patterns of each subject. One of the baseline shifts that appeared in the analog x-y plots of normal breathing was caused by the flushing of the spirometer with oxygen. Since

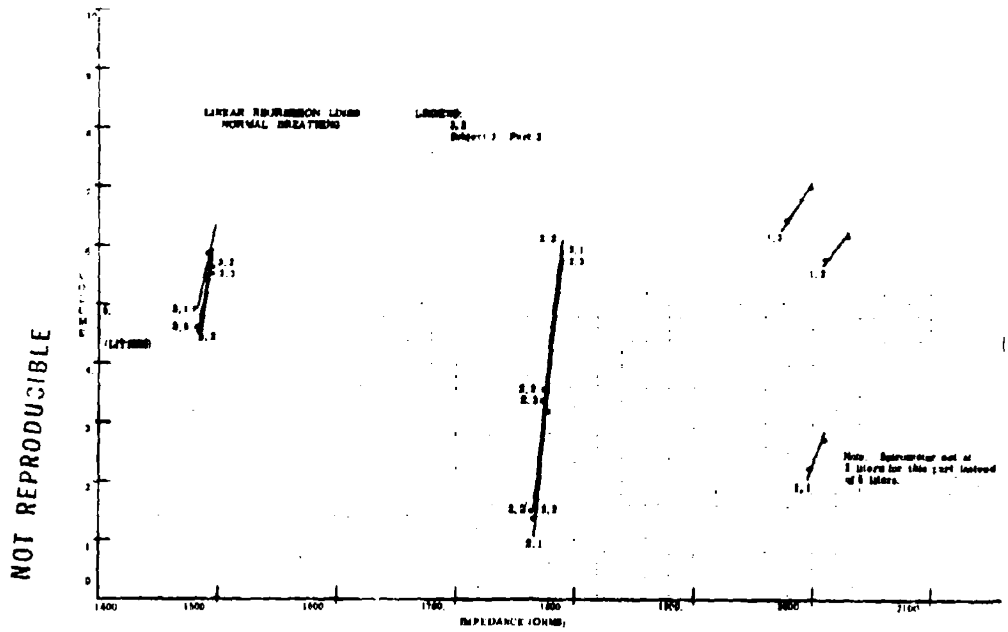


FIGURE 5

Linear regression lines for the normal breathing pattern of three subjects.

this change in spirometer output was not an indication of breathing, calculation of regression lines for the entire normal breathing pattern would yield false results. Therefore, the volume data were separated into blocks of similar baseline, and the linear regression lines were then calculated. These results are presented in figure 5. The identification procedure used in this graph (3,2; 3,3; etc.) stands for subject 3, part 2, and so on. Part 2 represents the second block of data that was separated according to baseline. Arrows are used on the linear regression lines to show the range of an average breath. The linear equations, standard errors of estimate, and correlation coefficients were calculated for each block of data and are listed in table III. The standard error of estimate is expressed in liters.

Figures 6 through 9 show digital x-y plots of various breathing sequences used in this experiment. The impedance waveforms were linearly converted from voltages to ohms, and the volume waveforms from voltages to liters. Figures 6 and 7 are plots of the maximum and minimum points for each breathing sequence. Figure 6 shows the scatter diagram for normal breathing of all three subjects, and figure 7 shows typical scatter diagrams for inhale and holding movements, impulse breathing, and electrode tapping. Figures 8 and 9 are plots of ΔV versus ΔZ, using the maximum and minimum points mentioned above. Figure 8 shows the scatter diagram of normal breathing for all three subjects, and figure 9 shows the typical scatter diagrams for arm movement, impulse breathing, and electrode tapping.

Table IV shows the comparison of slopes of normal breathing for linear regression analysis, maximum and minimum scatter diagrams, and ΔV versus ΔZ scatter diagrams. The effect of a 1% change in impedance is shown in liters for the various slopes considered in this table.

TABLE III

Statistical analysis of volume data by blocks of similar baseline

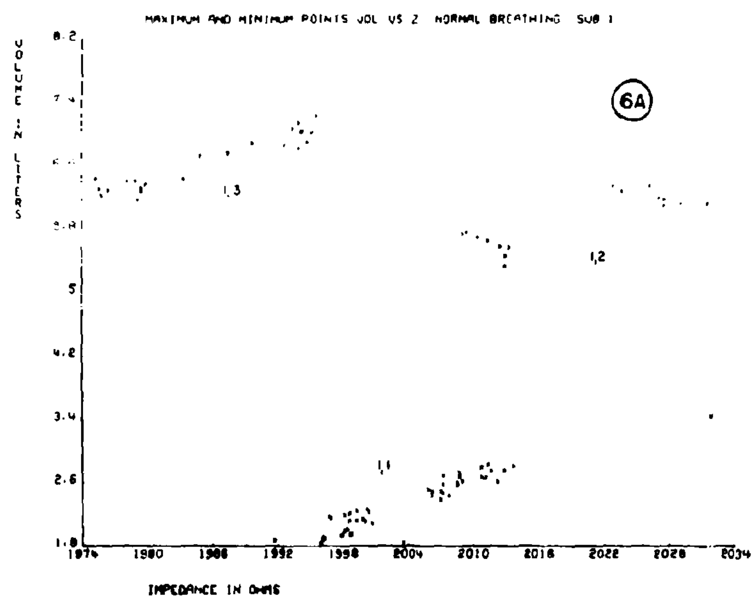
Trials	Linear Regression Equation $V = a_1 Z + a_0$		Standard Error of Estimate $S_{V,Z}$	Correlation Coefficient r	Number of Samples N
	a_1	a_0			
Sub 1 Part 1 (1, 1)	.05231	-102.3	.07114	0.9217	517
Sub 1 Part 2 (1, 2)	.02983	-54.26	1854	0.5820	312
Sub 1 Part 3 (1, 3)	.03347	-59.80	.1147	0.8112	508
Sub 2 Part 1 (2, 1)	.1865	-327.8	.3426	0.9441	794
Sub 2 Part 2 (2, 2)	.1916	-336.7	.2962	0.9505	585
Sub 2 Part 3 (2, 3)	.1808	-317.7	.3250	0.9478	254
Sub 3 Part 1 (3, 1)	.1023	-146.9	.1684	0.8541	405
Sub 3 Part 2 (3, 2)	.1211	-175.3	.1298	0.9204	450
Sub 3 Part 3 (3, 3)	.1192	-172.3	.1012	0.9527	364

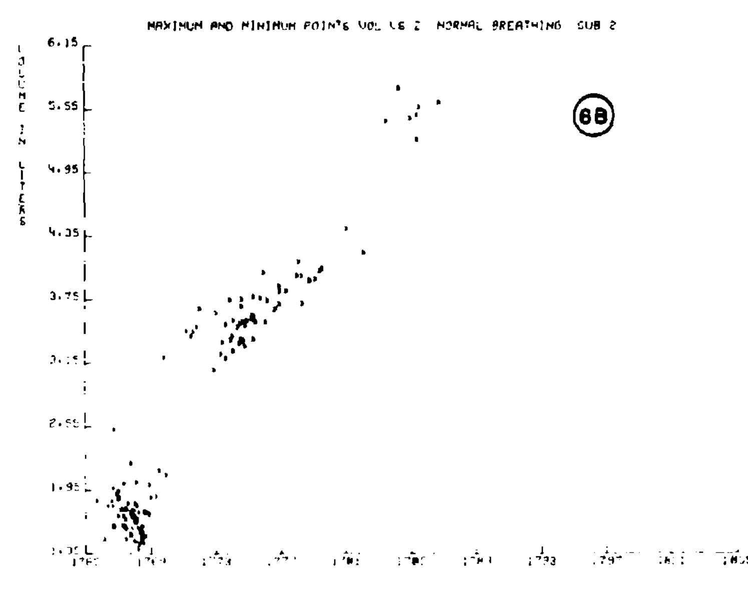
The volume error (expressed as a percent of standard) that would be obtained using a method other than linear regression to predict thoracic changes is also shown in table IV.

Table V shows the relationship between the slopes of normal breathing and artifact data for all three subjects. The slopes obtained for impulse breathing and electrode tapping are compared with the slopes obtained for normal breathing. The effect of a 1% change in impedance is shown in liters for the various slopes considered in this table. The volume error (expressed as a percent) shows the deviation obtained if the artifact data are used to predict normal breathing.

IV. DISCUSSION

The concept of measuring volume change by impedance was first introduced by Nyboer et al. in 1943 (43). Several investigators since then have reported on the relationship of tidal volume to thoracic impedance. In all cases their studies were hampered by movement artifacts. These artifacts can be classified as: (1) artifacts caused by changes in the body position, and (2) artifacts caused by the electrode-skin interface. Measurements by impedance are further complicated by the fact that the impedance signal obtained is different for each subject and each run. By use of the breathing pattern shown in table II, these artifacts were simulated in an exaggerated manner.


MAXIMUM AND MINIMUM POINTS VOL VS Z NORMAL BREATHING SUB 1
VOLUME IN LITERS
6A
1,3
1,2
1,1
IMPEDANCE IN OHMS


MAXIMUM AND MINIMUM POINTS VOL VS Z NORMAL BREATHING SUB 2
6B

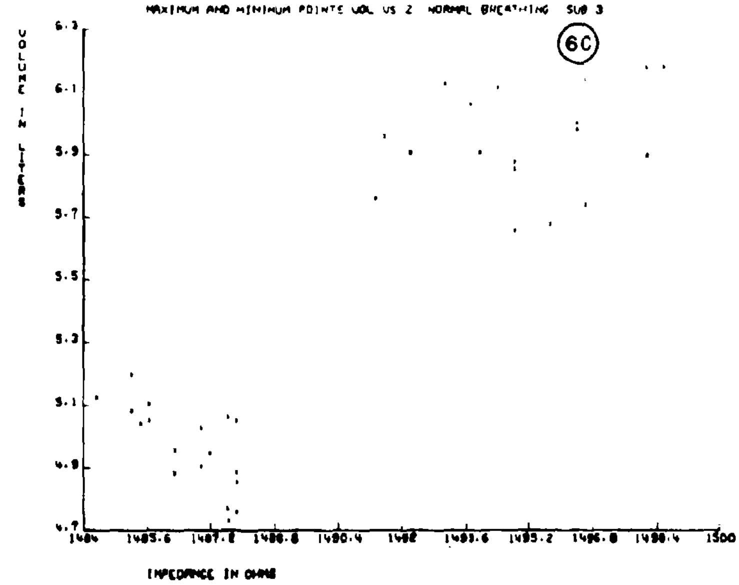
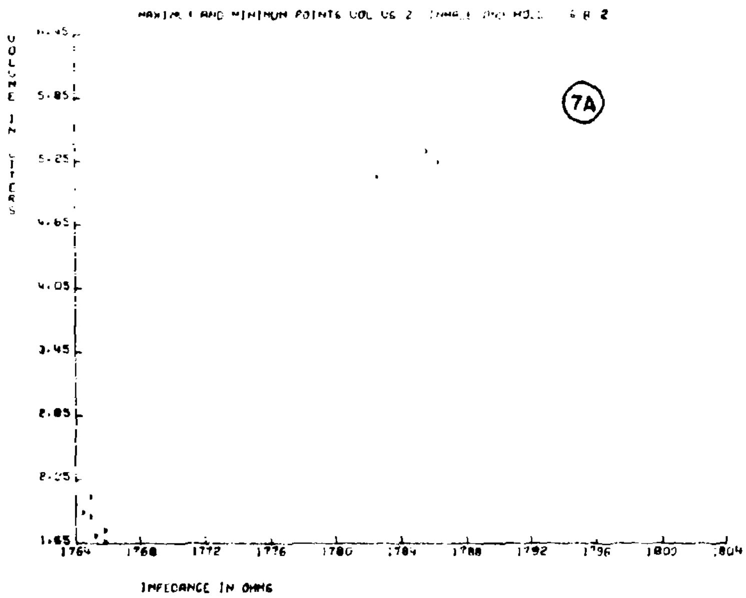
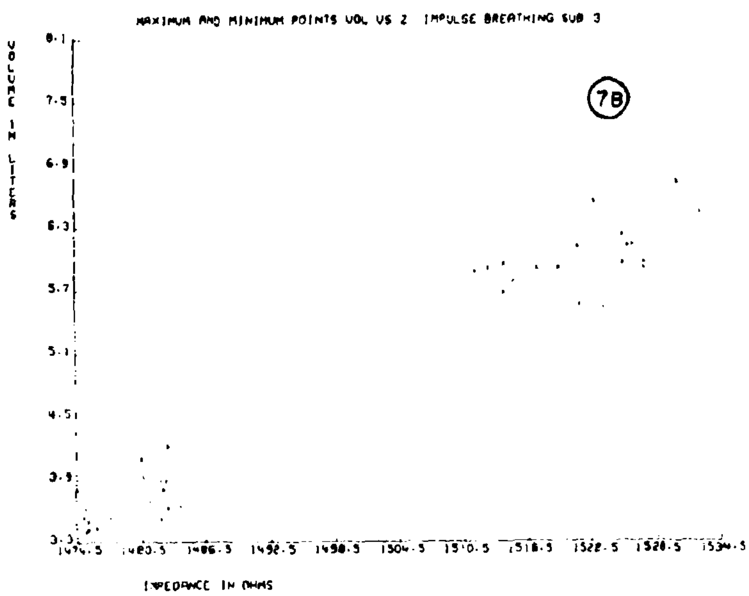


FIGURE 6

Scatter diagrams of maximum and minimum points for normal breathing in three subjects.



MAXIMUM AND MINIMUM POINTS VOL VS Z
7A
VOLUME IN LITERS
5.85
5.25
4.65
4.05
3.45
2.85
2.25
1.65
1764
1768
1772
1776
1780
1784
1788
1792
1796
1800
1804
IMPEDANCE IN OHMS



MAXIMUM AND MINIMUM POINTS VOL VS Z IMPULSE BREATHING SUB 3
7B
VOLUME IN LITERS
8.1
7.5
6.9
6.3
5.7
5.1
4.5
3.9
1480.5
1486.5
1492.5
1498.5
1504.5
1510.5
1516.5
1522.5
1528.5
1534.5
IMPEDANCE IN OHMS

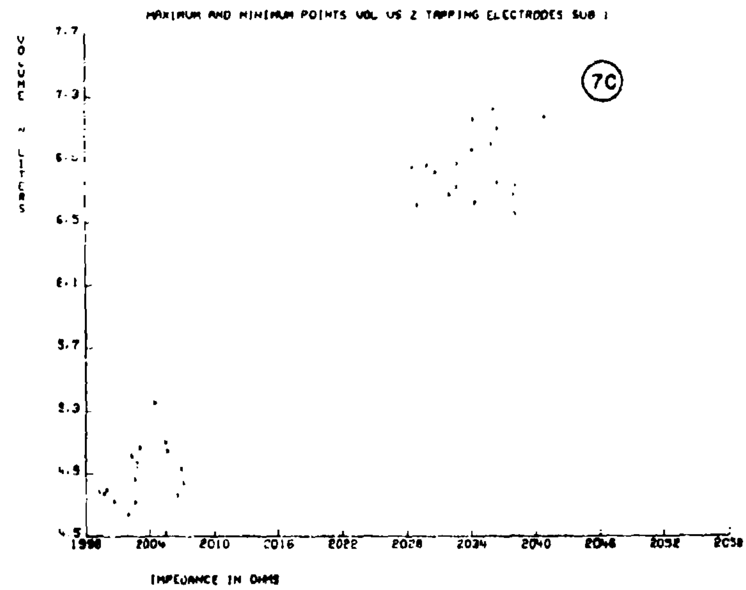
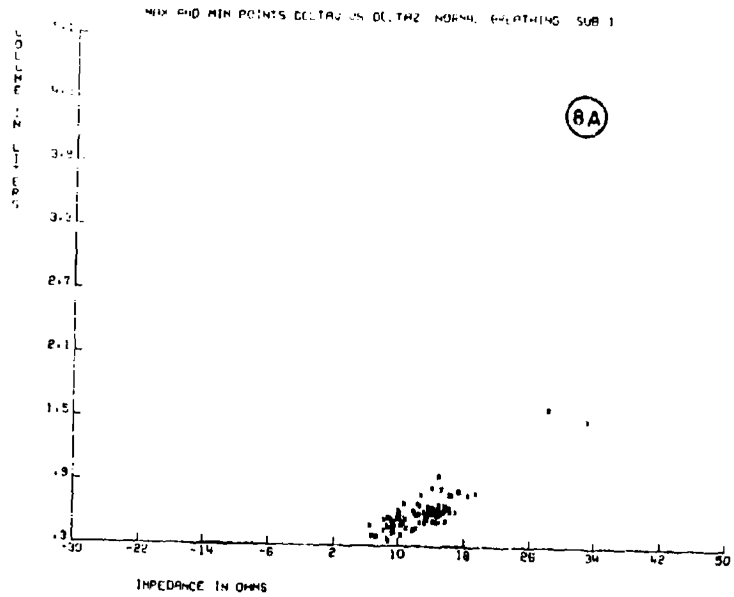
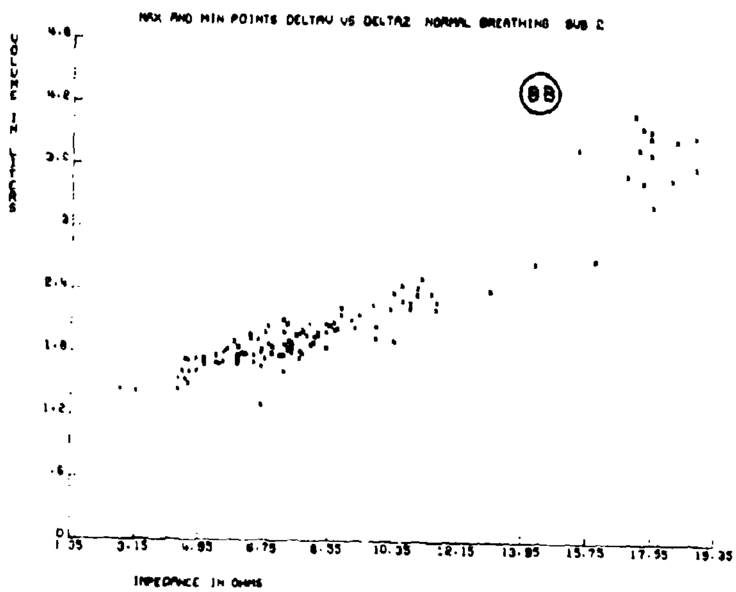


FIGURE 7

Scatter diagrams of maximum and minimum points for various breathing patterns: figure 7A—inhale and hold, subject 2; figure 7B—impulse breathing, subject 3; and figure 7C—electrode tapping, subject 1.



8A
IMPEDANCE IN OHMS



MAX AND MIN POINTS DELTAV VS DELTAZ NORMAL BREATHING SUB 2
8B
IMPEDANCE IN OHMS

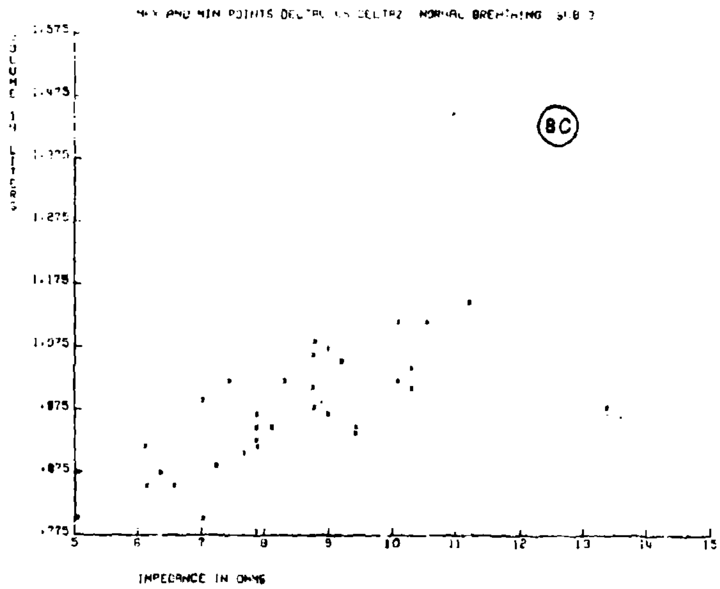
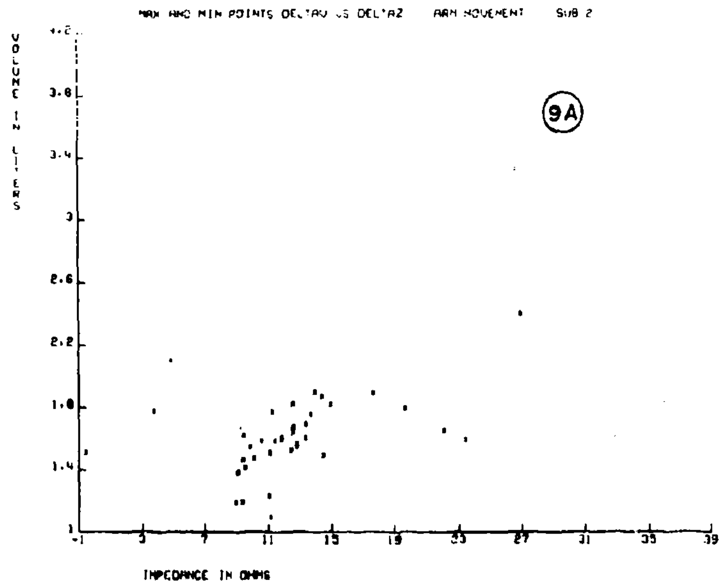
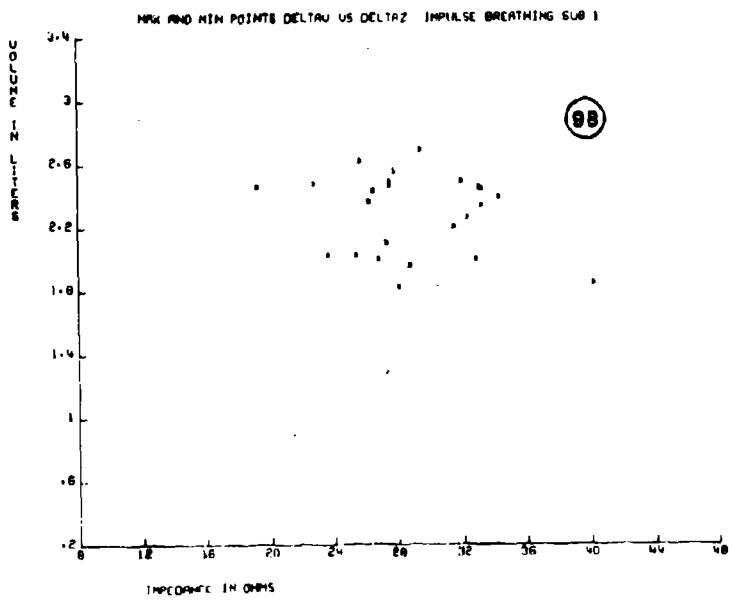


FIGURE 8

Scatter diagrams of ΔV versus ΔZ for normal breathing in three subjects.


MAX AND MIN POINTS DELTAV VS DELTAZ ARM MOVEMENT SUB 2
VOLUME IN LITERS
IMPEDANCE IN OHMS
9A


MAX AND MIN POINTS DELTAV VS DELTAZ IMPULSE BREATHING SUB 1
VOLUME IN LITERS
IMPEDANCE IN OHMS
9B

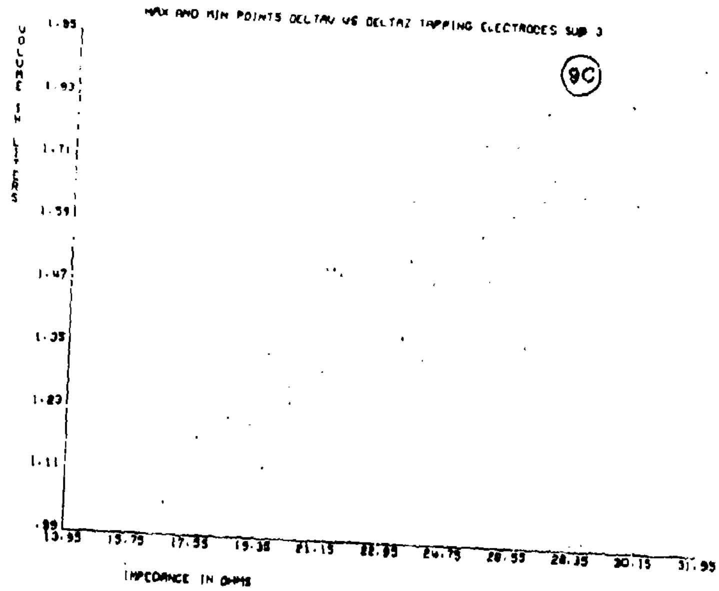


FIGURE 9

Scatter diagrams of ΔV versus ΔZ for various breathing patterns: figure 9A—arm movement, subject 3; figure 9B—impulse breathing, subject 1; and figure 9C—electrode tapping, subject 3.

TABLE IV

Comparison of slopes for the normal breathing data of three subjects

	SUB 1	SUB 2	SUB 3
(A) Linear regression slope for total waveform	.0385	.1863	.1142
(B) Linear regression slope for max. & min. points	.04156	.2152	.0989
(C) Average $\Delta V/\Delta Z$ slope for ΔV versus ΔZ plot	.04755	.24377	.11604
ΔV_A for 1%ΔZ	.73210	3.1671	1.5988
ΔV_B for 1%ΔZ	.78964	3.6584	1.3846
ΔV_C for 1%ΔZ	.90345	4.1441	1.62456
$\lvert \Delta V_A - \Delta V_B \rvert$	.05754	.4913	.2142
$\lvert \Delta V_A - \Delta V_C \rvert$	.17135	.9770	.02576
$\left[\frac{\Delta V_B - \Delta V_A}{\Delta V_A}\right]\%$	+7.8%	+15.5%	-13.4%
$\left[\frac{\Delta V_C - \Delta V_A}{\Delta V_A}\right]\%$	+19.0%	+23.5%	+1.6%

Two conclusions were made during the breathing pattern. The first was that each subject had a particular impedance range that remained constant during the entire experiment regardless of what simulated artifact was used. Second, even with a severe simulated artifact such as electrode tapping, the impedance signal could still be favorably related to the volume signal. Examples of these observations can be seen in the signal inserts of figure 4D. Hill et al. (24) suggest that a change in impedance level should be expected because of the effect of violent movements on the electrode-skin interface. Since the assumption of a constant range for each subject tends to contradict the results of Hill et al., a simple test was devised to study the impedance change at the electrode-skin interface. This test consisted of putting two electrodes together; first with electrode paste only, and then with paste and a damp sponge between the electrode surfaces. These were then tapped and the impedance change observed. In all tests, the impedance changed during the tapping and then returned to its original level. I conclude from this experiment that the electrode-skin interface is stable enough to return to its original level rather than be permanently altered by a violent movement. The assumption can then be made that a change in impedance level not directly attributable to a specific movement is probably caused by a change in lung volume. This is

TABLE V

Comparison of slopes between normal breathing and artifact data for three subjects

	Linear regression slope for max. & min. points			Average $\Delta V/\Delta Z$ slope for ΔV versus ΔZ plot		
	SUB 1	SUB 2	SUB 3	SUB 1	SUB 2	SUB 3
Normal	.04156	.21520	.09890	.04755	.24377	.11604
Tapping	.06215	.23610	.06242	.06384	.26165	.06340
Impulse	.07801	.13590	.05393	.07858	.10488	.05521
ΔV_N for 1% ΔZ	.78964	3.65840	1.38460	.90345	4.14409	1.62456
ΔV_T for 1% ΔZ	1.18085	4.01370	.87388	1.21296	4.44643	.88760
ΔV_I for 1% ΔZ	1.48219	2.31030	.75502	1.49302	1.78296	.77264
$\lvert \Delta V_N - \Delta V_T \rvert$	.39121	.35530	.51072	.30951	.30234	.73696
$\lvert \Delta V_N - \Delta V_I \rvert$	.69255	1.34810	.62958	.58957	2.36113	.85162
$\left[\frac{\Delta V_T - \Delta V_N}{\Delta V_N}\right]\%$	+49.5%	+9.71%	-36.8%	+34.2%	+7.29%	-45.3%
$\left[\frac{\Delta V_I - \Delta V_N}{\Delta V_N}\right]\%$	+87.7%	-36.8%	-45.4%	+65.2%	-56.9%	-52.4%

illustrated in the insert of figure 3B. The impedance level change after the sigh corresponds to a similar tidal volume change.

The fact that I could favorably relate the impedance signal to the volume signal in the presence of artifacts prompted an analysis by analog and digital data processing. The first procedure was to make analog x-y plots of the volume and impedance signal. This showed any trend in the data, thereby allowing determination of the order equation that would best fit the data. The difficulty in this method was the tendency of false spirometer outputs, ECG, and movement artifacts to mask or shift the visible trend.

Since it took approximately 20 minutes to complete the breathing sequence used in these studies, oxygen had to be fed into the spirometer. This forcing of oxygen into the spirometer caused a rapid shift in the output of the spirometer that could be falsely considered as a shift in lung volume. Unfortunately, this shift could not be analog filtered effectively because the frequency component of the shift was the same as the normal breathing frequency. Although digital technics could be used to correct this problem, a better control system to monitor and supply oxygen would eliminate it entirely.

The fundamental frequencies of the ECG are greater than 1 Hz. Therefore, the use of a low-pass filter with a cutoff of 1 Hz could eliminate the effect of ECG on the impedance waveform.

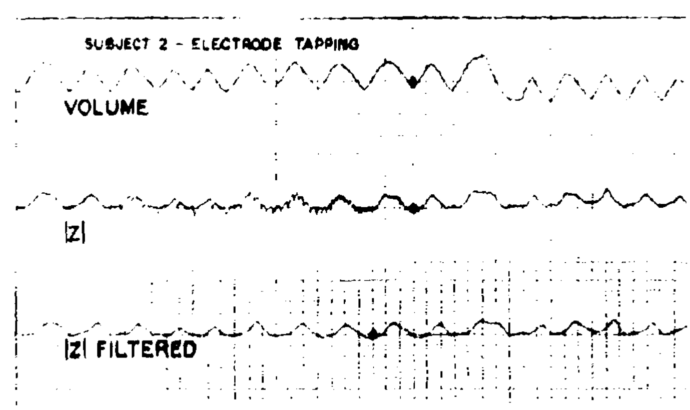


FIGURE 10

The use of analog filtering to eliminate the effects of electrode tapping on the impedance signal.

The third difficulty—movement artifact—had the greatest effect on the impedance waveform, as can be seen in figure 10 (electrode tapping). Movement artifacts that have frequencies sufficiently different from the normal breathing frequency can easily be analog filtered. Figure 10 also shows a sample of breathing that has been high-pass filtered at 1 Hz to eliminate the tapping effect. Movement artifacts that require more than simple filtering are discussed later in the text.

After a trend was established, the proper regression line (in this case linear) could be fitted to the data (fig. 3). The different level of volume for subject 1, part 1, was caused by the use of a value of 3 liters for the zero point of that run. In addition, the poor correlation coefficient of .58 for subject 1, part 2, can be explained by the large ECG artifact observed on this record.

The next analysis consisted of calculating and plotting the maximum and minimum points of both volume and impedance. It can be seen from figure 4 that a line drawn between the maximum and minimum point groupings would yield similar results to those obtained in the linear regression lines of figure 5. This procedure has the advantage of eliminating a large number of points and, as a result, much of the artifact. The fact that baseline shift is still recorded is one drawback to this procedure.

In many cases, the desired information is the relationship between the change in lung volume and the change in thoracic impedance. Therefore, the elimination of baseline shift is permissible. The baseline shift could be eliminated from the volume signal by a level detector circuit and an averaging circuit. The same result can also be obtained with digital programs. Another way of eliminating the effect of baseline shift is to plot ΔV versus ΔZ. A plot of ΔV versus ΔZ for normal breathing is shown in figure 8. The tight grouping (subject 1) and the linear groupings (subjects 2 and 3) show the linear similarity between the tidal volume and thoracic impedance. Similar graphs are shown in figure 9, for various simulated artifacts. The linear or grouping result is also evident in these examples even though the expanded scales tend to disperse the data. Therefore, a plot of ΔV versus ΔZ can be used to eliminate baseline shift and check the linearity between volume and impedance.

The solution for eliminating the baseline shift of the impedance signal is not as simple

as that used for eliminating the baseline shift in volume. The slowly varying changes of impedance, due to slow body movements or changing of the electrode-skin interface with time, alter the impedance signal related to breathing. Because these artifacts will mask the impedance signal and produce faulty results, they must be eliminated. Since the frequencies of these slow changes differ significantly from those of normal breathing, they can be filtered.

The faster changing artifacts caused by quicker body movements cannot be effectively filtered because of the similarities between their frequencies and those of normal breathing.

By using a computer, two possibilities exist for eliminating the baseline shift of the impedance signal. The use of precise numeric values gives the investigator a rigid limit within which he can accept data. The ability to store data allows the investigator to compare previous signals with the present signal and make decisions on whether artifact exists or not.

Signal averaging is another possible technic that could be used to get the desired change of impedance. This method will not work for short time periods, but it may work for the longer ones if the artifacts due to arm movement are random enough.

It should be noticed that the different displays represent a hierarchy of analysis—from strictly analog results to various digital programs. For a complete analysis of respiration, all these steps and more may be required; however, depending on the investigator's goal, only one or two displays may be sufficient. The computer also gives the investigator the advantage of getting precise relationships between tidal volume and thoracic impedance in liters and ohms rather than in volts as obtained in the analog solutions.

A comparison of slopes was made in order to evaluate the possibility of using the maximum and minimum point plots or ΔV versus ΔZ plots as methods of predicting tidal volume from thoracic impedance. The slopes for the maximum and minimum point plots were obtained by computing the slope of the linear regression line of volume on impedance for these data. The slopes for the ΔV versus ΔZ plots were obtained by computing the average value of ΔV and dividing by the average value of ΔZ.

The comparison shows that the maximum and minimum point method can be used with an expected error of $\pm$ 15% in predicting tidal volume from thoracic impedance and that the slope obtained from the ΔV versus ΔZ plot can be used to predict tidal volume with an expected error of $\pm$ 25%.

Slopes of normal data were compared with slopes of artifact data to evaluate the predictability of values for normal breathing under conditions of artifact. These results are shown in table V. For an artifact such as electrode tapping, an error of $\pm$ 50% can be expected in the prediction of values for normal breathing. This error is about the same for the maximum and minimum point method or for the ΔV versus ΔZ method. This table also shows that an error of 87% can be expected in predicting normal breathing values from impulse breathing. Although the error values shown for the ΔV versus ΔZ method are not as large as those obtained from the maximum and minimum method, they still represent significant error.

Several factors have to be considered in studying the larger error values shown in table V. The first of these is the use of maximum and minimum plots for normal breathing. This standard, as shown in table IV, can be off as much as 15%. The second factor is the effect of artifact on the impedance signal. Since this is a noisy signal, the linear regression line may not yield the best slope possible. It is also possible that the change in slope is due to a body impedance change rather than noise. In this case a higher error would be expected when trying to predict normal breathing from these data. An example of this is impulse breathing. As seen in table V, the error for predicting normal

breathing from impulse breathing is the largest for all three subjects regardless of how the slope was obtained. Since impulse breathing is more nonlinear than most artifacts (fig. 4A), this large error should be expected.

The last factor to consider is the amount of preconditioning performed on the data. The results shown in tables IV and V represent data that were not preconditioned. As seen in figure 10, the use of proper filtering would eliminate most of the noise and permit a much more accurate comparison.

In conclusion, two important points should be made: (1) The many factors that affect the impedance measured across the chest (16) cannot possibly be tested with 3 subjects. Many more studies, using a large number of subjects and conditions, are needed to confirm the trends indicated by this investigation. (2) A great many possibilities exist in the field of signal processing that have not been investigated. These technics will lead to new and better methods of presenting information; however, caution must be exercised to insure that methods which rely on past histories do not force new data to conform incorrectly to the old data.

V. CONCLUSIONS

The results of this study show a high degree of linear correlation between the tidal volume and thoracic impedance signals even under the conditions of induced artifact. In addition, with analog filtering, various digital technics, and a careful choice of the output display, most artifacts can be eliminated—leaving the investigator with a predictable output.

In view of these results, systems that eliminate many artifacts such as the constant current system suggested by Marsocci (38), the use of a tetrapolar electrode arrangement, and refined signal analysis should further insure the practicality of the impedance plethysmograph as a means of monitoring respiration in a space environment.

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13. ABSTRACT

The impedance plethysmograph was evaluated as a means of monitoring respiration of personnel in a space-flight environment. Emphasis was placed on studying the effects of limited movement (tapping of electrodes and skin stretching) on the impedance plethysmograph signal.

A bridge circuit and servospirometer were used to make extremely sensitive measurements of body impedance change and tidal volume. The linear relationship between tidal volume and thoracic impedance was determined by analog and digital methods. These methods were also used to investigate the amount of interference caused by various body movements on the impedance signal. Possible ways of eliminating error from such artifacts were also studied.

Very high linear correlation (r = .85 to .95) was obtained using a bridge circuit. The effects of artifact were reduced by the use of analog filtering, digital programming technics, and a careful choice of display.

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